In vivo thermal conductivity of the human forearm tissues

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DUCHARME, MICHEL B., AND PETER TIKUISIS. *In vivo thermal conductivity of the human forearm tissues.* J. Appl. Physiol. 70(6): 2682–2690, 1991.—The effective thermal conductivities of the skin + subcutaneous fat ($k_{eff\ skin+fat}$) and muscle ($k_{eff\ muscle}$) tissues of the human forearm at thermal steady state during immersion in water at temperatures (T_w) ranging from 15 to 36°C were determined. Tissue temperature (T_t) was continuously monitored by a calibrated multicouple probe during a 3-h immersion of the resting forearm. T_t was measured every 5 mm from the longitudinal axis of the forearm (determined from computed-tomography scanning) to the skin surface. Skin temperature (T_{sk}), heat loss ($\dot{H}_{sk}$), and blood flow ($\dot{Q}$) of the forearm, as well as rectal temperature (T_{re}) and arterial blood temperature at the brachial artery (T_{bla}), were measured during the experiments. When the k_{eff} values were calculated from the finite-element (FE) solution of the bioheat equation, $k_{eff\ skin+fat}$ ranged from 0.28 ± 0.03 to 0.73 ± 0.14 $W \cdot °C^{-1} \cdot m^{-1}$ and $k_{eff\ muscle}$ varied between 0.56 ± 0.05 and 1.91 ± 0.19 $W \cdot °C^{-1} \cdot m^{-1}$ from 15 to 36°C. The values of $k_{eff\ skin+fat}$ and $k_{eff\ muscle}$, calculated from the FE solution for $T_w \leq 30°C$, were not different from the average in vitro values obtained from the literature. The k_{eff} values of the forearm tissues were linearly related ($r = 0.80$, $P < 0.001$) to $\dot{Q}$ for $T_w \geq 30°C$. It was found that the muscle tissue could account for 92 ± 1% of the total forearm insulation during immersion in water between 15 and 36°C.

cold stress; intramuscular temperature; muscle tissue; thermal resistance; tissue insulation; water immersion

THE PHYSIOLOGICAL CONTROL of body insulation is an essential response to cold in humans (6). It is well recognized that, in addition to the cutaneous insulation provided by vasomotor adjustments, fat provides a significant insulative barrier to body heat loss during a cold thermal stress. Several authors have shown that resting individuals with a greater subcutaneous fat thickness cool at a slower rate than thin individuals (7, 22, 23). However, other studies have shown differences in thermoregulatory responses between individuals of similar adiposity (23) and similar thermal responses for individuals with different surface area-to-mass ratios (35). As suggested by several investigators (4, 7, 25, 32, 35), factors other than the subcutaneous fat thickness and the surface area-to-mass ratio affect the body cooling rate and/or the body insulation during cold stress. Thus, skeletal muscles have been repeatedly suggested to contribute to the body insulation (28, 32, 36), despite the evidence that the insulative value of the fatty tissue is two to three times greater than that of the muscle tissue per unit thickness (6, 21).

The insulation due to nonfatty tissue appears to be a cold-adaptative mechanism seen in Antarctic-acclimatized men (3), Korean diving women (32), and naked Australian aborigines (20). According to Hammel et al. (20), Australian aborigines with a fat thickness similar to their control subjects had tissue insulation values ~30% greater than those of the control group. However, several studies suggested that the cold-protective value of the nonfatty tissue insulation is not limited to cold-adapted subjects (7, 22, 23, 28, 35, 36). Carlson et al. (7) pointed out that for nonacclimatized subjects immersed in cold water the fraction of the body tissues contributing to the insulation was always greater than the estimated fat content. Later, Rennie et al. (32) reported that when the linear regression of the maximum body insulation to the subcutaneous fat thickness is extrapolated to zero fat thickness, a positive insulation value is found, which supports the existence of an in-series insulative muscle shell in addition to the subcutaneous fat. Furthermore, some investigators, either by graphical extrapolation (28) or by indirect determination (22, 23), estimated that as much as 80–90% of the maximum total body insulation (23, 28) or limb insulation (22) could be attributed to muscle tissue during cold stress. Despite all the above indirect evidence suggesting an important contribution of the muscle tissue to the total body insulation or limb insulation, no direct measurements are available in the literature. Furthermore, to our knowledge, only one study has investigated the in vivo or effective thermal conductivity of the muscle tissue, in this case for the lumbar region (31).

The purpose of the present study was to investigate the effect of a range of water temperatures on *1*) the effective thermal conductivity of the forearm tissues and *2*) the relative contribution of the forearm muscle tissue to the forearm tissue insulation. It is hypothesized that the forearm insulation is mainly achieved by the muscle tissue and is primarily dependent on the vasomotor status of the tissues.

MATERIAL AND METHODS

A detailed description of material and methods has been presented in another paper reporting the forearm tissue temperature at steady state during thermal stress in water (13). The following is a brief description of the methods used in this study.

Anthropometric measurements. Table 1 lists the mean anthropometric measurements performed on each of the 15 male subjects, aged 18–30 yr, that participated in this

TABLE 1. *Anthropometric characteristics of the subjects*

	Age, yr	Height, cm	Weight, kg	A_D, m^2	Forearm Circumference, cm	Forearm Radius, cm	Skin + Fat Thickness, cm	Muscle Thickness, cm	Proportion, %		
									Muscle	Skin + fat	Bone
Mean ± SE	22.5±1.1	180±1	74.2±2.4	1.9±0.03	27.5±0.6	4.1±0.1	0.30±0.02	3.8±0.1	78.0±1.2	13.8±1.1	8.2±0.3
Range	18–30	172–187	62.0–89.5	1.7–2.1	24.1–31.7	3.6–4.7	0.20–0.50	3.2–4.4	70.1–84.6	7.6–22.5	6.6–10.2

Anthropometric measurements were performed on both forearms and pooled for each subject. A_D, DuBois area.

study. The surface area of the subjects (A_D) was estimated with the formula of DuBois and DuBois (12). Anthropometric information of the forearm was measured from computed-tomography (CT) scans taken at the exact site of the muscle temperature measurements. The various characteristics were calculated using computerized planimetry (Graphic Tablet HP9111A, computer HP-85, Hewlett-Packard).

Temperature measurements. Tissue temperature (T_t) was continuously monitored during the experiments every 5 mm from the longitudinal axis of the forearm (determined by CT scan) to the skin surface on the bulkiest part of the forearm, ~9 cm distal from the olecranon process along the ulnar ridge. Temperatures were recorded with a fine calibrated multicouple probe (15) implanted under local anesthesia (skin refrigerant, ethyl chloride, Graham-Field) by use of a method described elsewhere (15). The position of the probe (and thus of each junction in the multicouple probe) in the forearm tissue was accurately known (±0.5 mm) and remained stable during the experiments. The T_t values were corrected for the thermal conductivity effect along the wires of the probe (15).

Skin temperature (T_{sk}) was measured a few millimeters from the site of the implanted probe with a calibrated 40-gauge type-T thermocouple. Rectal temperature (T_{re}) was measured using a calibrated thermistor probe inserted 15 cm beyond the anus. All temperature data were recorded continuously during the experiments by means of a computer-controlled data acquisition system (HP-3052A, computer HP-85, Hewlett-Packard). Mean values over 1-min periods were calculated for all measured variables.

The arterial blood temperature was recorded once before the immersion [$T_{bla(0)}$] at the distal portion of the brachial artery with a fine (40-gauge) calibrated type-T thermocouple probe inserted into a 20-gauge catheter. The arterial blood temperature at the end of the immersion (T_{bla}) was estimated from the $T_{bla(0)}$ value corrected for any fluctuation of T_{re} during the 3-h immersion and also for the effect of the countercurrent heat exchange occurring in the forearm as follows

$$T_{bla} = \{T_{bla(0)} + [T_{re(end)} - T_{re(0)}]\} + (0.075\ T_w - 2.700) \quad (1)$$

where $T_{re(end)}$ and $T_{re(0)}$ represent T_{re} at the end of the immersion and when $T_{bla(0)}$ was measured, respectively, and T_w is water temperature. The last component of *Eq. 1* (0.075 T_w − 2.700) represents the correction of $T_{bla(0)}$ for the countercurrent heat exchange, which was estimated from the study of Bazett et al. (Ref. 1; using data from Fig. 3 for forearm immersion at T_w = 19.3°C), assuming no effect at T_w = 36°C and a linear dependence with T_w. The first assumption is supported by the very low temperature differences expected in the forearm between the venous and arterial blood temperatures during water immersion at 36°C, and the second assumption is supported by the data of Bazett et al. (1) for air exposure. The puncture site for the measurement of T_{bla} was 5–6 cm proximal to the multicouple probe implantation site.

Blood flow measurements. The forearm blood flow ($\dot{Q}$) was measured by the plethysmograph Whitney gauge technique (38) using a mercury-in-Silastic circumference gauge (Parks Electronics Laboratory, Aloha, OR). The gauge was calibrated before each experiment by use of a previously described method (13). For each series of blood flow measurements, a mean value was calculated from five measurements over a 10-min period.

Heat flux measurements. The heat flux from the skin of the forearm ($\dot{H}_{sk}$) was continuously monitored during the experiments with two waterproofed heat flux transducers [HFTs; model HA13-18-10-P(C), Thermonetics, San Diego, CA] fixed to the skin with surgical tape (Blenderm, 3M, St. Paul, MN) on adjacent but opposite sides of the site of the multicouple probe implantation. Each HFT was recalibrated by a method described elsewhere (13, 16). Thermally conductive grease (Wakefield Engineering, Wakefield, MA) was applied under the HFTs to optimize the heat transfer between the skin and the device. A mean value of heat flux from the two HFTs was calculated over a 1-min period, and a correction was applied to that value for the effect of the thermal resistance of the HFT on the measured heat flux (16).

Experimental procedure. Subjects reported to the laboratory on two separate occasions within a 2-wk period. After the implantation of the multicouple probe, the subject, lightly dressed (T-shirt and casual pants), rested in a supine position under thermoneutral conditions (air temperature of 25.2 ± 1.1°C, relative humidity of 40 ± 2%) for 1 h. During that period, T_t, T_{sk}, and T_{re} were recorded continuously, the first measurement being performed ~10 min after the implantation of the probe. At the end of the resting period, the arterial blood temperature was recorded for 30 s, and the first series of blood flow measurements was performed with the subject sitting comfortably and resting the forearm under investigation on a surface at the level of the heart. The subject then immersed his forearm and hand for 3 h in a well-stirred water bath maintained at a constant temperature (T_w) of 15, 20, 30, 33, or 36°C (Haake F3 circulator temperature controller, Haake EK 51-1 immersion cooler, Haake E 52 circulator heater, Saddle Brook, NJ). At the end of the 3-h immersion, a second series of blood flow measurements was performed with the forearm still resting in the water bath. Each subject experienced two dif-

ferent temperatures chosen randomly, one for each forearm over a 2-wk period. A total of 30 experiments were performed, 6 experiments for each T_w condition.

Calculation of the effective thermal conductivity and insulation values. The effective thermal conductivity (k_{eff}) of the forearm tissues was calculated by using three different solutions: the nonconvective (NC) solution (33), the Pennes (PE) solution (29), and the finite-element (FE) solution (33). Each of the above solutions has specific assumptions and limitations, and the two latter are based on the bioheat equation developed by Pennes (29).

All three solutions were applied to the resting nonshivering forearm at thermal steady state assuming cylindrical geometry, negligible axial and tangential heat conduction, and uniform blood flow and thermal properties of the tissues. In addition, the NC solution assumed uniform metabolic heat production for the whole forearm (i.e., no Q_{10} effect) and no convective heat transfer between blood and tissue, with the implication that all the heat loss from the forearm originates from metabolic heat production. The calculation of k_{eff} (in $W \cdot °C^{-1} \cdot m^{-1}$) for the forearm tissues using the NC solution is expressed as (33)

$$k_{eff\ forearm} = \frac{0.5 r_{sk} \dot{H}_{sk}}{(T_0 - T_{sk})} \quad (2)$$

where r_{sk} is the forearm radius (in m) calculated from a photograph of the CT scan using computerized planimetry, $\dot{H}_{sk}$ is expressed in $W \cdot m^{-2}$, T_0 is the temperature at the longitudinal axis of the forearm (in °C), and T_{sk} is expressed in °C.

The PE solution also assumed uniform metabolic heat production but, in addition, included convective heat transfer between tissue and blood. The arterial blood temperature is fixed constant for the whole forearm, implying that the convective heat transfer between tissue and blood is limited to the capillary level. The values of $k_{eff\ forearm}$ were calculated using the PE solution of the bioheat equation (29). Inputs to both the NC and PE solutions are the temperatures at the skin and at the axis, the heat flux at the skin, and in the latter case the arterial blood temperature. It is important to note that the $k_{eff\ muscle}$ and $k_{eff\ skin+fat}$ values could not be calculated when the NC and PE solutions were used, because the assumptions for those solutions are applied to the whole forearm with no distinction for the different tissues.

The FE solution also involves the PE solution's assumptions but is applied to each 5-mm compartment inside the forearm. In addition, the FE solution applies the Q_{10} effect for the metabolic heat production, assumes uniform efficiency of convective heat exchange within each 5-mm compartment, and allows convective heat transfer to occur at each level of the vascular bed. This latter assumption leads to a decreasing temperature of the arterial blood as it flows from the axis to the skin. The calculation of k_{eff} using the FE solution (34) is expressed as

$$k_{eff\,[n]} = \frac{q_{n-1} r_{n-1} \ln (r_n / r_{n-1}) + (q_{v[n]}/2) \times [(r_n^2 - r_{n-1}^2)/2 - r_{n-1}^2 \ln (r_n / r_{n-1})]}{T_{n-1} - T_n} \quad (3)$$

where n is the compartment number, r_n is the outer radius at the compartment n, q_n is the heat flow at r_n (in $W \cdot m^{-2}$), q_{vn} is the sum of metabolic heat production and convective heat exchange for the compartment n (in $W \cdot m^{-3}$), and T_n is the tissue temperature at the outer boundary of the compartment n (in °C). The most superficial compartment of the forearm contained only subcutaneous fat and skin tissues because the boundaries were at the fascia and at the skin. The temperature at the fascia interface was obtained by extrapolating the muscle temperature to the radius corresponding to the fascia (calculated from a photograph of the CT scan using computerized planimetry). From *Eq. 3*, the following tissue conductivities were calculated as

$$k_{eff\ forearm} = \frac{\sum_{i=1}^{N} k_{eff\,[n]}}{N} \quad (4)$$

$$k_{eff\ muscle} = \frac{\sum_{i=1}^{N-1} k_{eff\,[n]}}{N-1} \quad (5)$$

$$k_{eff\ skin+fat} = k_{eff\,[N]} \quad (6)$$

where N is the number of compartments in the forearm.

The effective thermal insulation (R_{eff}) of the forearm tissues per unit length (in $°C \cdot W^{-1}$) was calculated for a cylindrical geometry by using the values from the FE solution as follows (33)

$$R_{eff\ forearm} = \sum_{i=1}^{N-1} \frac{\ln (r_{n+1}/r_n)}{k_{eff[n+1]} 2\pi L} \quad (7)$$

$$R_{eff\ muscle} = \sum_{i=1}^{N-2} \frac{\ln (r_{n+1}/r_n)}{k_{eff[n+1]} 2\pi L} \quad (8)$$

$$R_{eff\ skin+fat} = \frac{\ln (r_{sk}/r_{N-1})}{k_{eff\ skin+fat} 2\pi L} \quad (9)$$

where L is the length of the cylinder (in m), considered as unity.

Statistical analysis. The regression analysis, the tests of equality of lines (based on slope and intercept), and the independent and paired t tests were performed using the BMDP Statistical Program (Biomedical Computer Programs, BMDP, 1983, Los Angeles, CA). Data were analyzed by a one-factor analysis of variance to determine the difference between the values of $\dot{Q}$, $k_{eff\ forearm}$, $k_{eff\ muscle}$, $k_{eff\ skin+fat}$, $R_{eff\ forearm}$, $R_{eff\ muscle}$, and $R_{eff\ skin+fat}$ for each temperature tested. When the F ratio proved significant, the Newman-Keuls post hoc test was used to locate significant difference between the means. Where applicable, data are presented as means ± SE. The level of statistical significance was set at $P < 0.05$, unless otherwise stated.

RESULTS

No shivering activity was observed during any of the forearm immersions. T_{re} increased during the 3-h immersion period by 0.25 ± 0.03°C ($P < 0.001$) for all T_w conditions, probably because of the circadian cycle and/or tissue vasoconstriction in the extremities caused by the local cold stress. T_{re} measured at thermal steady state during the immersion averaged 37.45 ± 0.04°C for all subjects and T_w conditions tested.

Over the last 10 min of each 3-h immersion, the fore-

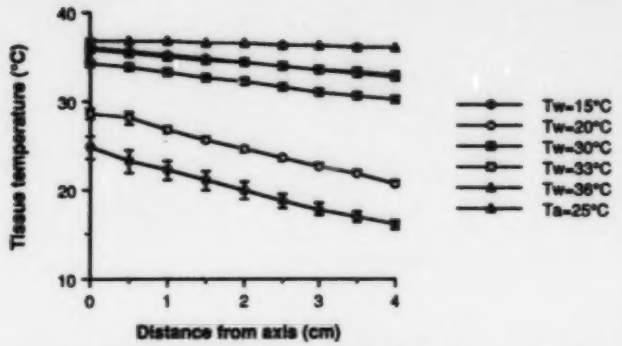

FIG. 1. Tissue temperature profiles inside forearm at thermal steady state for all temperature conditions tested. Values are means ± SE; $n = 6$ for each temperature condition. T_w, water temperature; T_a, ambient temperature.

arm tissue temperatures changed by only 0.03 ± 0.01°C and the forearm heat loss changed by only 1.18 ± 0.21 $W \cdot m^{-2}$. We therefore consider that the forearm had achieved a thermal steady state after 3 h of immersion at T_w = 15–36°C. All temperature and heat flow data are expressed as the mean values for the last 10 min of the 3-h immersion.

Temperature of the forearm tissues. Figure 1 depicts the tissue temperature profile of the forearm for each experimental condition. Although some interindividual variability was observed for the shape of the tissue temperature profile inside the forearm (some profiles presented slight convex or concave curvature) (14), there was a significant linear relationship at each temperature tested (r = 0.80–0.95, $P < 0.001$) between the forearm tissue temperature at steady state and the distance from the longitudinal axis of the forearm. The slope of these relationships decreased linearly with increasing T_w (r = 0.95, $P < 0.001$) from −2.13 ± 0.25°C · cm^{-1} at 15°C to −0.16 ± 0.02°C · cm^{-1} at 36°C. In each experimental condition, the maximal tissue temperature was measured at the longitudinal axis of the forearm. Furthermore the forearm tissue temperature at a given radial distance from the longitudinal axis (r) was linearly related to T_w (r = 0.96–0.99, $P < 0.001$), the slope of these relationships ranging from 0.57 ± 0.04°C · $°C^{-1}$ at r = 0.0 cm to 0.95 ± 0.02°C · $°C^{-1}$ at r = 4.0 cm.

Heat loss from the forearm. Figure 2 shows the linear relationship between T_w and the heat loss from the forearm at thermal steady state (slope = −2.97 ± 0.21, $P <$ 0.001). The forearm heat loss decreased almost fourfold, from an average of 95.4 ± 1.6 $W \cdot m^{-2}$ at 15°C to 26.0 ± 1.4 $W \cdot m^{-2}$ at 36°C.

Arterial blood temperature. The arterial blood temperature measured at the distal portion of the brachial artery before the immersion [$T_{bla(0)}$] averaged 36.62 ± 0.06°C, being 0.58 ± 0.03°C lower than the average T_m of 37.20 ± 0.04°C measured simultaneously ($P < 0.001$). After 3 h of immersion, the arterial blood temperature in the brachial artery (T_{bla} at the level of the proximal forearm) was calculated from *Eq. 1* to range from 35.20 ± 0.17°C at T_w = 15°C to 36.84 ± 0.12°C at T_w = 36°C. Although this represents the plausible range of T_{bla}, there is nevertheless a possibility that the arterial blood entering the forearm had a lower temperature than that estimated by *Eq. 1*. The consequence of a lower T_{bla} (down to ~27°C) on the k_{eff} values calculated by the FE model was observed, however, to be nonsignificant (≤5% change) (34).

Effective thermal conductivity of the forearm tissues. The different physiological values of T_t and H_{sk}, measured at thermal steady state, as well as the arterial blood temperature, were used to calculate $k_{eff\ forearm}$ by use of the three solutions previously described. Figure 3*A* depicts $k_{eff\ forearm}$ as a function of T_w. For the three solutions, $k_{eff\ forearm}$ calculated at thermal steady state during the immersions was minimal at T_w = 15–30°C, averaging 0.20 ± 0.01 $W \cdot °C^{-1} \cdot m^{-1}$ for NC, 0.16 ± 0.01 $W \cdot °C^{-1} \cdot m^{-1}$ for PE, and 0.53 ± 0.03 $W \cdot °C^{-1} \cdot m^{-1}$ for FE; there were no significant differences between the temperatures. The values of $k_{eff\ forearm}$ at 33°C (0.33 ± 0.03 $W \cdot °C^{-1} \cdot m^{-1}$ for NC, 0.25 ± 0.03 $W \cdot °C^{-1} \cdot m^{-1}$ for PE, and 0.81 ± 0.08 $W \cdot °C^{-1} \cdot m^{-1}$ for FE) were significantly greater than those measured at lower temperatures ($P < 0.01$) and significantly lower than those measured at 36°C ($P < 0.01$); (0.72 ± 0.08 $W \cdot °C^{-1} \cdot m^{-1}$ for NC, 0.60 ± 0.09 $W \cdot °C^{-1} \cdot m^{-1}$ for PE, and 1.72 ± 0.24 $W \cdot °C^{-1} \cdot m^{-1}$ for FE). For the entire range of T_w tested, the $k_{eff\ forearm}$ values calculated with the NC solution were not different from those calculated with the PE solution. However, the $k_{eff\ forearm}$ values calculated with both solutions were significantly lower than those for the FE solution ($P < 0.01$).

Figure 3*B* shows $k_{eff\ muscle}$ as a function of T_w, calculated by the FE solution. These values calculated at thermal steady state during immersion were minimal at T_w = 15–30°C, averaging 0.56 ± 0.03 $W \cdot °C^{-1} \cdot m^{-1}$; there was no significant difference between these temperatures. At 33°C, $k_{eff\ muscle}$ (0.75 ± 0.10 $W \cdot °C^{-1} \cdot m^{-1}$) was significantly greater ($P < 0.01$) than those calculated at lower temperatures and significantly lower ($P < 0.001$) than that calculated at 36°C (1.84 ± 0.19 $W \cdot °C^{-1} \cdot m^{-1}$). No values of $k_{eff\ muscle}$ were calculated using the NC and PE solutions because of the limitations of these solutions (see MATERIAL AND METHODS).

Figure 3*C* shows $k_{eff\ skin+fat}$ as a function of T_w, calculated by the FE solution. These values calculated at thermal steady state during immersion were minimal at T_w = 15–30°C, averaging 0.29 ± 0.02 $W \cdot °C^{-1} \cdot m^{-1}$; there were no significant differences between these temperatures. The $k_{eff\ skin+fat}$ were significantly greater ($P < 0.01$) at 33

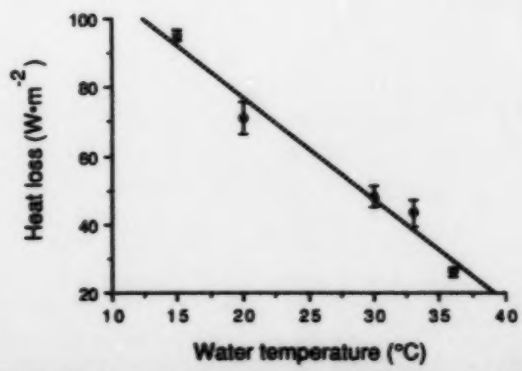

FIG. 2. Heat loss from forearm at thermal steady state for all temperature conditions tested. Values are means ± SE; $n = 6$ for each temperature condition.

and 36°C than at lower temperatures. No values of $k_{eff\ skin+fat}$ were calculated for the NC and PE solutions because of the limitation of the solutions (see MATERIAL AND METHODS).

Forearm blood flow. $\dot{Q}$ measured at thermal steady-state during the immersion was minimal at T_w ranging from 15 (0.78 ± 0.15 ml · min^{-1} · 100 ml tissue^{-1}) to 30°C (1.33 ± 0.24 ml · min^{-1} · 100 ml tissue^{-1}). There was no significant difference between these $\dot{Q}$ values. $\dot{Q}$ at 33°C (2.94 ± 0.17 ml · min^{-1} · 100 ml tissue^{-1}) was significantly greater ($P < 0.001$) than that measured at lower temperatures and significantly lower ($P < 0.001$) than that measured at 36°C (6.20 ± 0.51 ml · min^{-1} · 100 ml tissue^{-1}).

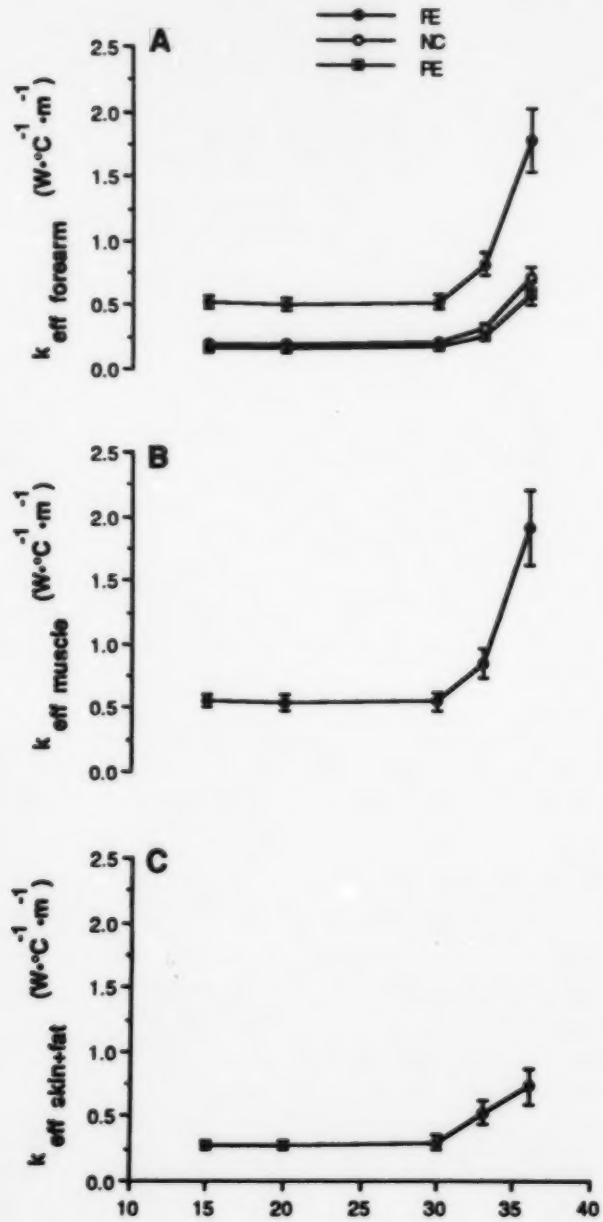

FIG. 3. Effective thermal conductivity (k_{eff}) of forearm tissues calculated at thermal steady state from 3 different solutions for all temperature conditions tested: for the whole forearm (*A*), for muscle tissue (*B*), and for skin + subcutaneous fat tissues (*C*). Values are means ± SE; $n = 6$ for each temperature condition, except skin + subcutaneous fat tissues, where $n = 5$ at 15 and 30°C. FE, finite element; NC, nonconvective; PE, Pennes solution.

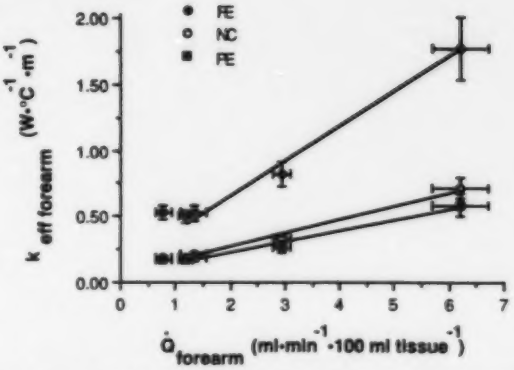

FIG. 4. Effective thermal conductivity of forearm tissues calculated at thermal steady state from 3 different solutions as a function of forearm blood flow ($\dot{Q}$). Values are means ± SE; $n = 6$ for each temperature condition.

For all three solutions used to calculate $k_{eff\ forearm}$, there was a significant linear relationship ($P < 0.001$) between $k_{eff\ forearm}$ and $\dot{Q}$ at thermal steady state for $T_w \geq 30$°C ($r = 0.79$ for NC, $r = 0.68$ for PE, and $r = 0.80$ for FE), $k_{eff\ forearm}$ values increasing with increasing $\dot{Q}$ (Fig. 4). For $T_w = 15$–30°C, no significant relationship was found between $k_{eff\ forearm}$ and $\dot{Q}$ for the three solutions. The relationships between $k_{eff\ forearm}$ and $\dot{Q}$ calculated from the NC and PE solutions were not different. However, both relationships were significantly different from that calculated from the FE solution ($P < 0.001$).

Thermal insulation of the forearm tissues. Figure 5 depicts $R_{eff\ forearm}$, $R_{eff\ muscle}$, and $R_{eff\ skin+fat}$, calculated from the FE solution, as a function of T_w. In addition, Fig. 5 shows the in vitro thermal insulation of the same tissues calculated from the average value of in vitro thermal conductivity obtained from the literature (8, 30) and from the average tissue thickness from our subjects. $R_{eff\ forearm}$, $R_{eff\ muscle}$, and $R_{eff\ skin+fat}$ calculated at thermal steady state during the immersions were maximum at $T_w = 15$–30°C, with no significant difference between temperatures. For the whole range of T_w tested, $R_{eff\ muscle}$ was responsible for 92 ± 1% of $R_{eff\ forearm}$, with no significant difference between T_w.

DISCUSSION

The thermal conductivity is a thermophysical property of the tissue that depends on its structure and composition (i.e., protein, fat, water, and ash content) and, to a lesser extent, its temperature. Among the components of the biological tissues, fatty tissue has the lowest thermal conductivity (0.19 W · °C^{-1} · m^{-1}) (33) and water has the highest (0.63 W · °C^{-1} · m^{-1}) (33). Thus, the in vitro thermal conductivity value of biological materials is expected to fall within these limits. Some authors have developed equations to calculate the thermal conductivity of tissues

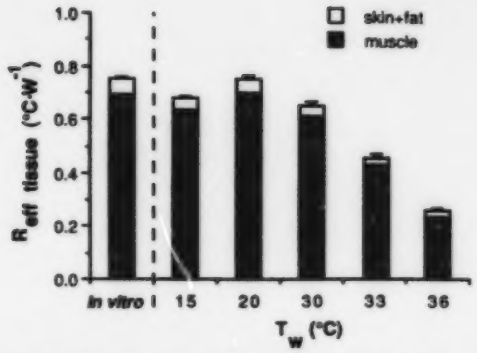

FIG. 5. Effective thermal insulation (R_{eff}) of skin + subcutaneous fat and muscle tissues of forearm calculated at thermal steady state from FE solution for all temperature conditions tested. Values are means ± SE; n = 6 for each temperature condition.

from their composition to within 5% of their in vitro values (11, 30). In living biological systems, however, the thermal conductivity of tissue is also dependent on the convective heat transfer by the blood, thus giving a higher "apparent" thermal conductivity to the living tissue. This apparent or in vivo thermal conductivity of tissue has been defined by Burton and Bazett (5) as an "effective" thermal conductivity, which is a practical index of the rate at which heat flows through tissue by the combined means of direct conduction and convective transport of heat by blood flow. When the status of the tissue in vivo approaches the in vitro conditions (i.e., absence or very low values of blood flow), some studies have shown that k_{eff} approaches the in vitro thermal conductivity of the tissue (10, 11, 31, 36). Therefore the lowest limit of k_{eff} expected for a specific living tissue is its in vitro value, and the highest limit will be determined by the maximum blood flow. Keller and Seiler (24) have determined from theoretical considerations that k_{eff} does not vary by more than a factor of 4.

Various techniques have been developed to calculate k_{eff} of living tissues. Although several of these are limited to the superficial tissues (27, 37), some can be applied to deep tissues, such as the thermal conductivity probe technique (2) and the internal calorimetry technique (19), which use active heated probes, the passive probe technique (10), which is limited to one instantaneous measurement for each probe implantation, and the passive temperature profile technique (29). The latest technique was used in the present study because this was the only one available to determine simultaneously the k_{eff} of all tissues of a limb, as opposed to a point determination.

Calculation of the effective thermal conductivity of the forearm tissues. As previously described, three different solutions were used to calculate $k_{eff\ forearm}$ from the steady-state tissue temperature profile measurements and the heat loss values. For the range of T_w tested, the values of $k_{eff\ forearm}$ were similar for the NC and PE solutions but lower than the values obtained with the FE solution (see Fig. 3). From the literature, the average value of in vitro thermal conductivity for the human and animal muscle tissues is 0.48 ± 0.06 W · °C^{-1} · m^{-1}, while that for the skin + fat tissues averages 0.23 ± 0.02 W · °C^{-1} · m^{-1} (8, 31). These values are not different from the $k_{eff\ muscle}$ and $k_{eff\ skin+fat}$ values obtained with the FE solution at T_w = 15–30°C. Because $\dot{Q}$ was very low at T_w = 15–30°C (1.14 ± 0.11 ml · min^{-1} · ml tissue^{-1}) due to tissue vasoconstriction, it was therefore expected, on the basis of previous studies on unperfused or cooled tissues (10, 11, 31, 36) that $k_{eff\ muscle}$ and $k_{eff\ skin+fat}$ values similar to the in vitro values would be obtained. This is supported by previous studies that have shown minimum thermal conductivity of tissue similar to the in vitro values during immersion at $T_w \leq 30$°C (4, 36). Although no $k_{eff\ muscle}$ and $k_{eff\ skin+fat}$ values are available for the NC and PE solutions, it can be expected from the $k_{eff\ forearm}$ obtained with these two solutions that the $k_{eff\ muscle}$ and $k_{eff\ skin+fat}$ values would be lower than the in vitro values.

These findings, taken together with the fact that the forearm tissue temperature profiles estimated from the NC and PE solutions overestimated the measured temperatures profiles (34), suggest that the FE solution gives the most reasonable values of $k_{eff\ forearm}$. This implies that the convective heat transfer between blood and tissue is an important component of the heat exchange occurring in the forearm and that this heat exchange is not limited to the capillary level. It was estimated by the FE solution that the majority of the convective heat exchange between the blood and the forearm tissues occurred outside the capillary level with no difference between the T_w tested. This is consistent with the findings of Lemons et al. (26), who suggested that the thermally significant vessels are three or more generations from the capillary level. The FE solution predicted that the contribution of the convective heat transfer between blood and tissue to the forearm heat loss decreased with the increase of T_w, from 95% at 15°C to 92% at 20°C, 77% at 30°C, 68% at 33°C, and 41% at 36°C, the balance originating from the metabolic heat production of the tissue. This is attributed to the decrease of the temperature difference between the arterial blood and the forearm tissues and to the increase of the metabolic heat production in the tissues caused by the Q_{10} effect as T_w increased.

Blood flow and effective thermal conductivity of tissues. Several authors have observed for the skin (6, 24, 27), the liver (2), and the kidney tissues (20) that the k_{eff} values were directly related to $\dot{Q}$. Similarly, a positive linear relationship between $k_{eff\ forearm}$ and $\dot{Q}$ was observed in this study for T_w = 30–36°C. For that range of T_w, an increase in $\dot{Q}$ of 4 ml · min^{-1} · 100 ml tissue^{-1} yielded a conductivity increment ($k_{eff} - k_{in\ vitro}$) of 1 W · °C^{-1} · m^{-1} for the forearm tissues. In addition, for a range of $\dot{Q}$ between 1 and 6 ml · min^{-1} · 100 ml tissue^{-1}, the $k_{eff\ forearm}$ increased more than threefold. Maximum values of $k_{eff\ forearm}$ were probably not achieved in the present study, because, according to the studies of Burton and Bazett (5) and Keller and Seiler (24), the k_{eff} of the tissue can change to up four- to sixfold in response to the variation of the vascular tone. This implies, however, that the linear relationship observed between the k_{eff} values and $\dot{Q}$ in the present study will flatten for high values of $\dot{Q}$. For T_w < 30°C or for $\dot{Q}$ below ~1 ml · min^{-1} · 100 ml tissue^{-1}, no relationship was observed between $k_{eff\ forearm}$ and $\dot{Q}$, likely because of a fixed number of open capillaries, whereas for T_w = 30–36°C, the number of open capillaries likely increases linearly with $\dot{Q}$.

Figure 3, *B* and *C*, shows that for $T_w > 30°C$, the values of $k_{eff\ muscle}$ and $k_{eff\ skin+fat}$ increased significantly with T_w, as was reported for $k_{eff\ forearm}$ values. It is expected from the linear relationship of $k_{eff\ forearm}$ with $\dot{Q}$ that, with an increase of $T_w > 30°C$, blood flow would increase for the skin + fat and muscle tissues, probably because of the release of the vasoconstriction tone in both tissues that was present during water immersion at $T_w \leq 30°C$. This assumption is supported by Cooper et al. (9), who reported, using the epinephrine iontophoresis method, a linear positive relationship between the skin and the muscle blood flow for forearm blood flow between 1 and 6 $ml \cdot min^{-1} \cdot ml\ tissue^{-1}$.

Effective thermal conductivity of muscle and skin + fat tissues. The $k_{eff\ muscle}$ values calculated from the FE solution ranged between $0.56 \pm 0.03\ W \cdot °C^{-1} \cdot m^{-1}$ at $T_w \leq 30°C$ and $1.84 \pm 0.19\ W \cdot °C^{-1} \cdot m^{-1}$ at 36°C, a variation of more than threefold for the range of T_w tested (see Fig. 3). These values of $k_{eff\ forearm}$ are similar to those observed for the muscle tissues of the lumbar region by Reader (31), who reported $k_{eff\ muscle}$ values between 0.53 and 1.96 $W \cdot °C^{-1} \cdot m^{-1}$ when the dorsal region was locally exposed to a cold or a warm environment for 2 h. The same author observed, similarly to our study, that with progressive cooling the $k_{eff\ muscle}$ decreased to a minimum value that stayed constant even with further decrease of the local environmental temperature. The same type of comparison can also be done for $k_{eff\ skin+fat}$ values, which ranged for the present study between $0.29 \pm 0.02\ W \cdot °C^{-1} \cdot m^{-1}$ at $T_w \leq 30°C$ and $0.73 \pm 0.14\ W \cdot °C^{-1} \cdot m^{-1}$ at 36°C, compared with values ranging from $0.31\ W \cdot °C^{-1} \cdot m^{-1}$ for cold tissues to $0.42\ W \cdot °C^{-1} \cdot m^{-1}$ for warm tissues for Reader's study (31), $0.23\ W \cdot °C^{-1} \cdot m^{-1}$ for the study of Veicsteinas et al. (36) for the thigh during water immersion at $T_w = 30°C$, and $0.26\ W \cdot °C^{-1} \cdot m^{-1}$ for the study of Ferretti et al. (18) for the thigh during water immersion at an average $T_w = 28°C$. It seems therefore that, when cooled, the skin + fat and muscle tissues of the forearm achieved k_{eff} values similar to those for the lumbar region and the thigh, which suggests that, for a similar thermal stress, the k_{eff} of a specific tissue may be independent of body region.

The interindividual variation in k_{eff} for the different forearm tissues can be attributed to the variation in the tissue composition between subjects (mainly in fat content) and/or a variation in the vasomotor tone for the same thermal stress.

Calculation of forearm effective thermal conductivity with the linear form of Fourier's law of conduction. The k_{eff} of various tissues of the body is occasionally calculated by some authors from the linear form of Fourier's law of conduction (22) stated as

$$k_{eff} = \frac{\dot{H}_{sk} \cdot \Delta x}{T_{re} - T_{sk}} \qquad (10)$$

where $\dot{H}_{sk}$ is the heat loss from the skin of the body segment, Δx is the thickness of the tissue, and T_{re} and T_{sk} are, respectively, the core and skin temperatures of the body segment. T_{re} is usually assumed to equal the core temperature of the segment (T_c), despite the fact that studies have shown that this assumption is not applicable for the limbs (13, 29). When $k_{eff\ forearm}$ values are calculated with the linear form of Fourier's law of conduction, assuming T_c to be equal to the T_{re} of the subject, the values ranged from an average of $0.20 \pm 0.01\ W \cdot °C^{-1} \cdot m^{-1}$ at T_w = 15–30°C to $0.40 \pm 0.03\ W \cdot °C^{-1} \cdot m^{-1}$ at 33°C and $0.81 \pm 0.09\ W \cdot °C^{-1} \cdot m^{-1}$ at 36°C. Compared with the $k_{eff\ forearm}$ values calculated from the FE solution, the $k_{eff\ forearm}$ obtained from *Eq. 10* were underestimated by approximately twofold for the whole range of T_w tested. This is attributed to *1*) the assumption that the T_c of the segment has a value equal to T_{re} and *2*) the use of the linear form of Fourier's law of conduction, which is not applicable for limbs (i.e., cylindrical segments) where there is internal heat production within the segment. Therefore the calculation of the in vivo thermal conductivity and thermal insulation of limb tissues by use of the linear form of Fourier's equation leads to a significant error (underestimation for the conductivity values and overestimation for the insulation values). However, when T_c is measured and the linear form of Fourier's equation is used for the calculation of superficial or deep tissue k_{eff} having a plane geometry or a low degree of curvature (as in the case for the superficial tissue of a limb), the solution is closer to the FE solution (~20% deviation). This is supported by the studies of Reader (31), Veicsteinas et al. (36), and Ferretti et al. (18), who reported values of k_{eff} similar to those measured by the FE solution for similar thermal stress conditions.

The assumptions inherent in the linear form of Fourier's equation are violated when the equation is used to determine the thermal conductance of tissue where internal heat production is present. The error involved worsens as the metabolic rate increases and as the convective heat exchange between blood and tissue increases, which can occur with an increasing blood-tissue temperature difference and/or increasing blood flow. Although blood flow delivers heat to tissue via convection and thereby contributes to the internal heat liberated in the tissue (34), its flow through tissue enhances the movement of heat within the tissue. This local convective transport of heat is one of the two mechanisms of heat flow in tissue; the other is conduction driven by the temperature difference within the tissue; both mechanisms are inherent to the k_{eff} values. It must be emphasized that the local convective transport of heat within tissue via blood flow is separate from the convective heat delivered to tissue by the blood (34).

Recently, Ferretti et al. (18) applied the linear form of Fourier's equation to determine the local convective component of heat loss (q_c) from the skin + fat of the human thigh under exercise conditions during cold stress

$$q_c = \dot{H} - K_{eff\ min}(T_{sf} - T_{sk}) \qquad (11)$$

where $\dot{H}$ is the measured heat flux at the skin surface, T_{sf} is the temperature at a fixed distance below the skin surface, and $K_{eff\ min}$ is the effective conductance (in $W \cdot °C^{-1} \cdot m^{-2}$) determined during a rest condition in a cool environment where it was assumed that the convective heat delivered to tissue by the blood was negligible. In essence, $K_{eff\ min}$ was applied to a temperature difference ($T_{sf} - T_{sk}$) measured during exercise under the as-

sumption that $K_{eff\,min}$ is unaffected by changes in blood flow. Yet, studies have shown that changing the tissue water content affects significantly the conductivity of the tissue (11, 30). In that case, an increase of the tissue blood flow will increase the $K_{eff\,min}$ value, because the composition of the tissue is altered by an increase of the tissue blood content having higher thermal conductivity than the tissue itself. Ignoring such an increase will tend to overestimate the convective component of the heat loss as determined by Ferretti et al. (18).

Effective thermal insulation of the forearm tissues. The calculation of the R_{eff} values for the forearm tissues by use of the FE solution have allowed a determination of the contribution of the muscle insulation to the total forearm insulation. The data show that 92 ± 1% of the forearm tissues' insulation was attributed to the skeletal muscle for T_w = 15–36°C. This observation is in agreement with the estimations made by several authors using graphical extrapolation or indirect determination that, during a cold stress, the resting nonshivering muscle tissue could contribute 80–90% of the overall tissue insulation (22, 23, 28, 36). The present study extends this observation to neutral conditions in water. Furthermore, as discussed earlier, if the thermal properties of skin + fat and muscle tissues are similar for different parts of the body, it could be assumed that for the whole body the majority of the total body insulation may be attributed to the muscle tissue. However, further investigation is needed to verify this hypothesis.

In conclusion, this study shows that, at thermal steady state during water immersion, k_{eff} values calculated by the FE solution for the skin + subcutaneous fat and muscle tissues are minimum at $T_w \leq 30$°C and not different from the in vitro values. Above 30°C, the values for $k_{eff\,forearm}$ are linearly related to $\dot{Q}$. Furthermore, for the whole range of T_w tested, 92% of the effective forearm insulation is attributed to the skeletal muscle tissue. This suggests that the skeletal muscle is the main tissue responsible for the forearm insulation during cold stress and at neutral conditions in nonshivering resting humans.

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REFERENCES

1. BAZETT, H. C., L. LOVE, M. NEWTON, AND C. EISENBERG. Temperature changes in blood flowing in arteries and veins in man. *J. Appl. Physiol.* 1: 3–19, 1948.
2. BOWMAN, H. F. Thermodynamics of tissue heating: modeling and measurements for temperature distributions. *Med. Phys. Monogr.* 8: 511–548, 1982.
3. BUDD, G. M. Effects of cold exposure and exercise in a wet, cold antarctic climate. *J. Appl. Physiol.* 20: 417–422, 1965.
4. BULLARD, R. W., AND G. M. RAPP. Problems of body heat loss in water immersion. *Aerosp. Med.* 41: 1269–1277, 1970.
5. BURTON, A. C., AND H. C. BAZETT. A study of the average temperature of the tissues, of the exchange of heat and vasomotor responses in man by means of a bath calorimeter. *Am. J. Physiol.* 117: 36–53, 1936.
6. BURTON, A. C., AND O. G. EDHOLM. *Man in a Cold Environment.* London: Arnold, 1955.
7. CARLSON, L. D., A. C. L. HSICH, F. FULLINGTON, AND R. W. ELSNER. Immersion in cold water and body tissue insulation. *J. Aviat. Med.* 29: 145–152, 1958.
8. CHATO, J. C. A survey of thermal conductivity and diffusivity data on biological materials. *J. Biomed. Eng.* 66-WA/HT-37: 1–9, 1967.
9. COOPER, K. E., O. G. EDHOLM, AND R. F. MOTTRAM. The blood flow in skin and muscle of the human forearm. *J. Physiol. Lond.* 128: 258–267, 1955.
10. COOPER, T. E., AND G. J. TREZEK. A probe technique for determining the thermal conductivity of tissue. *J. Biomed. Eng.* 70-WA/HT-18: 1–12, 1970.
11. COOPER, T. E., AND G. J. TREZEK. Correlation of thermal properties of some human tissue with water content. *Aerosp. Med.* 42: 24–27, 1971.
12. DUBOIS, D., AND E. F. DUBOIS. A formula to estimate the approximate surface area if height and weight be known. *Arch. Intern. Med.* 17: 863–871, 1916.
13. DUCHARME, M. B. *Effective Insulative Properties of the Human Forearm Tissues* (PhD thesis). Toronto, Ontario: University of Toronto, 1990.
14. DUCHARME, M. B. *Individual Variability of Tissue Temperature Profile in the Human Forearm During Water Immersion.* North York, Ontario: Defence Civil Inst. Environ. Med., 1991. (DCIEM Rep. 91–10)
15. DUCHARME, M. B., AND J. FRIM. A multicouple probe for temperature gradient measurements in biological materials. *J. Appl. Physiol.* 65: 2337–2342, 1988.
16. DUCHARME, M. B., J. FRIM, AND P. TIKUISIS. Errors in heat flux measurements due to the thermal resistance of heat flux discs. *J. Appl. Physiol.* 69: 776–784, 1990.
17. FERRETTI, G., A. VEICSTENIAS, AND D. W. RENNIE. Conductive and convective heat flows of exercising humans in cold water. *J. Appl. Physiol.* 67: 2473–2480, 1989.
18. GRAYSON, J. Internal calorimetry in the determination of thermal conductivity and blood flow. *J. Physiol. Lond.* 118: 54–72, 1952.
19. HAMMEL, H. T., R. W. ELSNER, D. H. LEMESSURIER, H. T. ANDERSON, AND F. A. MILAN. Thermal and metabolic responses of the Australian aborigine exposed to moderate cold in summer. *J. Appl. Physiol.* 14: 605–615, 1959.
20. HATFIELD, H. S., AND L. G. C. PUGH. Thermal conductivity of human fat and muscle. *Nature Lond.* 168: 918–919, 1951.
21. HAYWARD, M. G., AND W. R. KEATINGE. Roles of subcutaneous fat and thermoregulatory reflexes in determining ability to stabilize body temperature in water. *J. Physiol. Lond.* 320: 229–251, 1981.
22. JEQUIER, E., P.-H. GYGAX, P. PITTER, AND A. VANOTTI. Increased thermal body insulation: relationship to the development of obesity. *J. Appl. Physiol.* 36: 674–678, 1974.
23. KELLER, K. H., AND L. SEILER, JR. An analysis of peripheral heat transfer in man. *J. Appl. Physiol.* 30: 779–786, 1971.
24. KOLLIAS, J., L. BARTLETT, V. BERGSTEINOVA, J. S. SKINNER, E. R. BUSKIRK, AND W. C. NICHOLAS. Metabolic and thermal responses of women during cooling in water. *J. Appl. Physiol.* 36: 577–580, 1974.
25. LEMONS, D. E., S. CHIEN, L. I. CRAWSHAW, S. WEINBAUM, AND L. M. JIJI. Significance of vessel size and type in vascular heat transfer. *Am. J. Physiol.* 253 (*Regulatory Integrative Comp. Physiol.* 22): R128–R135, 1987.
26. LIPKIN, M., AND J. D. HARDY. Measurement of some thermal properties of human tissues. *J. Appl. Physiol.* 7: 212–217, 1954.
27. PARK, Y. S., D. R. PENDERGAST, AND D. W. RENNIE. Decrease in body insulation with exercise in cool water. *Undersea Biomed. Res.* 11: 159–168, 1984.
28. PENNES, H. H. Analysis of tissue and arterial blood temperatures in the resting human forearm. *J. Appl. Physiol.* 1: 93–122, 1948.
29. POPPENDIEK, H. F., R. RANDALL, J. A. BREEDEN, J. E. CHAMBERS, AND J. R. MURPHY. Thermal conductivity measurements and pre-

dictions for biological fluids and tissues. *Cryobiology* 3: 318–327, 1966.
31. READER, S. R. The effective thermal conductivity of normal and rheumatic tissues in response to cooling. *Clin. Sci. Lond.* 11: 1–12, 1952.
32. RENNIE, D. W., B. G. COVINO, B. J. HOWELL, S. H. HONG, B. S. KANG, AND S. K. HONG. Physical insulation of Korean diving women. *J. Appl. Physiol.* 17: 961–966, 1962.
33. SEKINS, K. M., AND A. F. EMERY. Thermal science for physical medicine. In: *Therapeutic Heat and Cold*, edited by J. F. Lehmann and J. V. Basrajian. Baltimore, MD: Williams & Wilkins, p. 91–99, 1982.
34. TIKUISIS, P., AND M. B. DUCHARME. Finite-element model solution of thermal conductivity of muscle during cold water immersion. *J. Appl. Physiol.* 70: 2673–2681, 1991.
35. TONER, M. M., M. N. SAWKA, M. E. FOLEY, AND K. B. PANDOLF. Effects of body mass and morphology on thermal responses in water. *J. Appl. Physiol.* 60: 521–525, 1986.
36. VEICSTEINAS, A., G. FERRETTI, AND D. W. RENNIE. Superficial shell insulation in resting and exercising men in cold water. *J. Appl. Physiol.* 52: 1557–1564, 1982.
37. VENDRIK, A. J., AND J. J. VOS. A method for the measurement of the thermal conductivity of the human skin. *J. Appl. Physiol.* 11: 211–215, 1957.
38. WHITNEY, R. J. The measurement of volume changes in human limbs. *J. Physiol. Lond.* 121: 1–27, 1953.